CONTROL OF UPPER-LIMB PROSTHESES IN SEVERAL DEGREES OF FREEDOM

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1. INTRODUCTION

This presentation deals with the state of the project of controlling an artificial upper-limb prosthesis in several degrees of freedom, conducted at Colorado State University, as of June 1974. It concerns work on the design and construction of a controller to control the above prosthesis by means of microelectronics circuitry and of microprocessor computational hardware. This presentation will describe the progress on two alternative aspects of the project which are also complementary to one another, since their combination can be employed to extend the number of limb functions to be controlled. These two aspects are: first, the toe-actuated controller and second, the EMG-actuated controller. Whereas the toe-actuated controller is already operational in our laboratory, in hardware, the EMG-actuated controller is presently in a simulation stage. Each of these controllers is presently a three-function limb-controller. However, both are in principle extendable to a higher number of limb functions. Furthermore, as stated earlier, both designs can be combined in a design having some limb functions controlled via the toes and others via myoelectric signals. This could be the case when both arms are amputated above the elbow, or when one foot is also not functional. Obviously, in cases where the amputee cannot use any toes, only EMG control will be considered.

We comment that the following research was started in November 1971, under a VA contract, and work on the EMG approach was initiated in July 1973, though some work on this aspect of the project was started already in January 1973. Involved in this research were, Dr. D. Graupe, Dr. W. K. Cline (until June 1974 on a part-time basis when he was a candidate and graduate assistant in our laboratory), Mr. T.K. Kaplon, M.S. (until January 1974 when he was an M.S. candidate and graduate assistant in our laboratory), and Mr. W. J. Monlux, B.S., since January 1974 (presently an M.S. candidate and graduate assistant in our laboratory). Mr. A.A.M. Beex, M.S., has joined our laboratory in July 1974, as a graduate assistant and Ph.D. candidate.

2. TOE-ACTUATED CONTROL

The toe-actuated controller designed at Colorado State University is now operational in our laboratory. It facilitates control of three limb functions, namely, elbow flexion, wrist rotation, and prehension. Continuous control of both position and speed or torque is possible up to prescribed limits, in either positive or negative directions, via pulse-width modulation. In contrast to an earlier toe-actuated control design, by S. Alderson (1) (1954), where three toe movements were required per limb function, the present design (2), employing logic techniques and microelectronic hardware unavailable in 1954, requires only one toe movement. Furthermore, the present design reduces to a minimum the complexity of actuation, as far as the amputee is concerned, and eliminates by parity-check logic computations and inhibitions, some most likely actuation errors.

Specifically, the toe-actuated controller employs actuation of soft (resistive) strain gages by means of the big and the little toe of each foot, noting that the above toes are noninteracting in their movement. Dead-zones in the strain-gage amplifier circuitry further avoid actuation by accidental toe pressure. Also, the system is inhibited during walking, thus facilitating control only while sitting, lying, or standing.

The strain gage signal is fed to the control logic which is in terms of a microelectronic integrated circuit. The latter coordinates the limb functions according to Table 1 [a] and computes inhibitions and parity checks for errors in actuation. (Modifications to this Table for cases of bilateral amputees, for amputees who have only the use of one foot, and for simultaneous actuation of two different limb functions have also been designed and are given in the appendices. Subsequently, the output of the control logic is fed to the appropriate permanent magnet d.c. motors to execute the control of the three limb functions considered, namely, elbow flexion, wrist rotation, and prehension. Hence, continuous position and speed or torque control up to a limit are facilitated at the positive and negative direction for each of the above limb functions (apart from the wrist rotation function, where speed control was not considered to be important).

[a] Little toe movement can be replaced by a back movement (down) of the big toe that is different from BRD and BLD of Table 1, or by a movement of any one or more of the toes excluding the big toe, according to the amputee's convenience.

We comment that the above system is now complete in hardware form and includes the logic controller, the arm mechanisms, rechargeable Ni-Cad batteries, linkages, motors, interfaces, and special sandals with their toe-actuation strain gages, etc. The motors draw a maximum of 1 amp. at 18 v.d.c., the controller and actuators together taking only 3.5 mW. The whole system is packaged in a 5×3×1½ in. box, excluding the batteries and the sandals with their strain gages, whereas the motors are housed in the arm itself. We also comment that there is no reason why other functions than those mentioned could not be exercised instead. Furthermore, noting Table 1, an extension to four limb functions is possible.

TABLE 1.—*Command Functions for Controller*

		Flex elbow			Extend elbow			Close grasp			Open grasp			Rotate wrist	
		F	N	S	F	N	S	F	N	S	F	N	S	CW	CCW
BR	U	X	X	X	—	—	—	DX	—	—	—	—	DX	—	—
	D	—	—	—	X	X	X	—	—	DX	DX	—	—	—	—
BL	U	DX	—	—	—	—	DX	X	X	X	—	—	—	—	
	D	—	—	DX	DX	—	—	—	—	—	X	X	X	—	—
LR	D	—	—	—	—	—	—	—	—	—	—	—	—	X	—
LL	D	—	—	—	—	—	—	—	—	—	—	—	—	—	X

Key:

BR: big right toe
BL: big left toe
LR: little right toe
LL: little left toe
U: (press) up
D: (press) down

N: normal speed
F: increase speed (or torque, force)
S: decrease speed (or torque, force)
X: press (close switch)
DX: press after delay (such that X precedes DX)
CW: clockwise
CCW: counterclockwise

3. EMG-ACTUATED SUBSYSTEM

Whereas the system of Section 2 employs toe movements as input actuation signals to the prosthesis controller, the controller is not limited to just this kind of input. Also, not in all cases will that kind of input be available, since cases of injury or paralysis of one or more toes or feet may occur. The natural input actuation sources for control at will are thus myoelectric signals.

Major problems in multifunctional prosthesis control via EMG signals are filtering the EMG signals from noise (environmental, EKG, etc.) and from other interacting EMG signals, as well as the problem of recognizing (separating) the EMG signals related to different limb functions. For adequate and reliable solution of such filtering and recognition, a rigorous statistical analysis of EMG signals is required. Unfortunately, the EMG analysis in the literature is mostly restricted to ad hoc methods of spectral density and correlation evaluations (3, 4, 5, 6, 7, 8, 9, 10, 11). Hence EMG prosthesis control was essentially of single-function nature, and the resulting multifunctional control was thus always of ad hoc nature, employing amplitude-level coding (12). This in turn has usually required level-training of the amputee to actuate multifunctional control to affect reliability and amputee's comfort. Methods of pattern recognition of EMG signals were recently proposed (13). However, these have involved a large number of electrodes and complex memory and computation regarding ad hoc and usually predetermined tasks to again affect reliability, simplicity, comfort, and speed. Noting the recent advances in applying rigorous time-series analysis techniques to EEG analysis (14, 15, 16), a similar but non ad hoc philosophy has been proposed by this author for application to EMG signals (17, 18, 19, 20). These advances, coupled with the enormous advances in microcomputer and microprocessor hardware, form the basis for this work and for its reliability with today's hardware.

The present approach to EMG analysis recognizes the nonstationary nonlinear nature of the EMG signals. Furthermore, it recognizes the practical constraints in any realization of filtering and identification or recognition, when employed in conjunction with a realistic prosthesis, the constraints being those of time available for processing and of cost and weight of computational hardware to be carried by the amputee (say, in his pocket). These constraints must and do certainly lead to compromises and to simplifying assumptions in any analysis, that will detract from the rigor of the analysis. However, via using microcomputer hardware, an adequate and reliable solution based on rigorous analysis is still realizable as long as the number of degrees of freedom considered is small. (An incorporation of the latter system with limb control as in Section 2 (2, 18) thus leads to considerable prosthesis maneuverability via EMG-actuation, while causing hardly any mental strain to the amputee in terms of a need to learn complex actuation combinations, etc.)

The present analysis has been based on the nature of time-series. Their pattern can be parameterized into a finite set of parameters of a linear signal model (16), which forms a reduced minimal set as compared with the (almost) infinitum of values of the pattern itself. These parameters need not all be stationary or unique per each function for function separability. However, if at least some of these parameters will be such that their range of variation for a given limb function does not overlap with that of other functions, then such separation will be possible; this was found to be the case in all our surface-electrode EMG recordings (over 250 records).

Specifically, noting the constraints above, we have decided to examine minimum-order autoregressive-moving-average (ARMA) models of stationary time-series (16, 21, 22, 23, 24, 25, 26, 27). Although the EMG signal is not a stationary one, we have shown (see Table 2 and References 18, 19, and 20) that the signal is sufficiently stationary per each of the limb functions considered, to yield ARMA parameters whose range of variation with time is adequately small to facilitate discrimination of these limb functions (elbow flexion, wrist rotation, and prehension). The choice of a minimum-parameter linear ARMA model is thus justified by its yielding the required function discrimination and noting that the identification of a nonlinear or of a nonstationary model is impossible in practice, especially since for practical actuation of limb functions, the identification and the recognition of a function must be completed within about 0.1 second.

Although in the case of Table 2, separation is already possible when considering only one electrode location, this may and need not be the general case. However, if we wish to resolve different limb functions, we may generally have to record EMG signals at several (few) locations, such that *at some of these locations at least some of the parameters of the various functions* do not change in time so much that these functions become indistinguishable. Further results obtained have all facilitated similar function separation.

TABLE 2.—*ARMA Model for Recorded EMG—Ranges of Parameters*
$s=10$; Electrode at Biceps

	$\hat{\phi}_1$	$\hat{\phi}_2$	$\hat{\theta}_1$	$\hat{\theta}_2$	σ_w^2 ($\times 10^{-3}$)
L1, F1	$-.8923\pm.1120$	$-.0809\pm.1230$	$-.0700\pm.0880$	$-.0817\pm.0631$	$4.388\pm.261$
L1, F2	$-1.801\pm.0480$	$+.8517\pm.0160$	$-.4635\pm.210$	$-.2023\pm.025$	$1.213\pm.265$
L1, F3	$-1.906\pm.010$	$+.9354\pm.013$	$-.0303\pm.290$	$+.3111\pm.016$	$2.152\pm.690$

Key: F = Limb Function, L = Electrode Location, σ_w^2 = Variance of model residual, s = Order of Autoregressive Model, ϕ_i = Autoregressive Parameters, θ_i = Moving Average Parameters

Results based on 10-bit EMG data

We comment that the parameters considered above need not be similar for equivalent functions of different patients. To overcome this problem, complete offline identification (to establish parameter ranges) must be made for each patient prior to connecting his prothesis-controller to the identification and filtering microcomputer that is to execute the recognition.

Obviously, since external or biological non-EMG noise (say noise due to motors, fluorescent lights, etc.) may have overlapping parameters, this should be filtered first. The latter filtering is presently accomplished by using an optimal-linear Kalman filter based on the ARMA parameters above as identified from the recorded EMG data without any a priori knowledge of ad hoc assumptions. We note that EKG interactions may be filtered out by employing special low-band EKG filters.

Whereas all our results have been obtained from processing EMG data on the Colorado State University CDC 6400 computer, we have also completed a preliminary design of the microcomputer hardware required for on-amputee computation and control. Furthermore, a simulation study based on using INTEL 8008-1 and 8080 microprocessors is underway, employing a double length word (16 bits in total), noting the fixed point arithmetic features of that hardware. This preliminary study indicated that with presently available INTEL eight-bit fixed point hardware and double length words, all computation can be completed within 0.1 seconds. This is in fact within our real time needs. However, noting that fixed point arithmetic reduces somewhat the accuracy, a floating point system is certainly most desirable. Since INTEL and other companies have already announced a 16-bit microprocessor equivalent to the present INTEL 8080, but which is 16 times faster (1.25 μs instruction time), an incorporation of a floating point routine in this microprocessor hardware is possible within the constraint on computation time. Hence, our goal of 0.1 second total time, with accuracy as achieved on the CDC 6400 computer for our data from the 10-bit analog-digital converter (see Table 2), will be met when this hardware is marketed in early 1975. Furthermore, since our INTEL simulation program is usable to directly program the microprocessor hardware (namely, to produce hardware as required for our purpose), and noting that very similar if not fully identical programs can be used for the faster microprocessors, our work should not only establish the feasibility of our approach considering the constraints on cost, speed, performance, and weight that are obvious in real prosthesis hardware, but also should actually produce this prosthesis-borne EMG processing hardware.

4. CONCLUSIONS

The work that has been described concerns a multi-functional artificial upper-limb controller. The design was based on actuation via toe inputs and via EMG inputs. Whereas the toe-actuated system is presently complete and merely requires on-amputee testing and the related modifications, the second actuation system is only at a design stage. However, even the latter has already been proved to be feasible for realistic prosthesis application by processing of real data with microcomputer hardware, and it awaits computation and adjustments of false alarm probabilities, and of concrete hardware realization (with hardware to be available in early 1975 but which is in principle similar to existing, though somewhat slower, hardware). Consequently, a multi-functional arm using the controller of Section 1. with either toe or EMG actuation or both, for controlling three to six limb functions, should be complete by the end of the present project.

We cannot help emphasizing the importance of the recent progress in stochastic filtering and estimation software and theory and, above all, in microcomputer hardware, to achieve this end. We believe that microcomputer hardware must and will play a major role in artificial limbs and organs and in bioengineering diagnostic and surgical systems, due to its enormous computational power encapsulated in small and cheap hardware. This should be of tremendous benefit to the disabled and to the sick, opening new avenues in treatment, diagnosis, and in artificial and semi-artificial limbs and organs.

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APPENDIX I—DESIGN MODIFICATION FOR CASES OF DOUBLE AMPUTATION (BILATERAL ABOVE-ELBOW AMPUTATION)

In this modification, the design is modified to allow elbow flexion, wrist rotation and prehension of each arm in both directions (up or down or clockwise and counterclockwise), as in Table 2, to facilitate actuation of two upper-limb prostheses in cases of bilateral amputees.

Table 1

FUNCTION		RIGHT ARM	LEFT ARM
ELBOW	FLEX	BRU	BLU
	EXTEND	BRD	BLD
PREHENSION	OPEN	LRU	LLU
	CLOSE	LRD	LLD
WRIST ROTATION	CW	BRU AND BLU	LRU AND LLU
	CCW	BRD AND BLD	LRD AND LLD

KEY:
BRU - BIG RIGHT TOE UP
BRD - BIG RIGHT TOE DOWN
BLU - BIG LEFT TOE UP
BLD - BIG LEFT TOE DOWN
LRU - LITTLE RIGHT TOE UP
LRD - LITTLE RIGHT TOE DOWN
LLU - LITTLE LEFT TOE UP
LLD - LITTLE LEFT TOE DOWN
CW - CLOCKWISE
CCW - COUNTERCLOCKWISE

Comments

In this design a total of six functions are controlled (in two directions each).
No speed control is facilitated.
Control is continuous via pulse width modulation.
All other toe combinations than in table above are inhibited to disallow errors in actuation.
Cases of one above- and one below-elbow amputation can be accommodated via small (simplifying) modifications in the design.
Simultaneous use of right and left arm is possible.
Simultaneous actuation of elbow flexion and prehension and of wrist and prehension is possible.

APPENDIX II—DESIGN MODIFICATION FOR CASES OF ONE-FOOT-DISABILITY

In this modification only one foot can be used, such that an upper-limb amputee who cannot use one of his feet can still actuate the toe-controlled upper-limb prosthesis.

TABLE 2

FUNCTION		TOE ACTUATION
ELBOW FLEXION	FLEX	BU
	EXTEND	BD
PREHENSION	OPEN	LU
	CLOSE	LD
WRIST ROTATION	CW	BU AND LU
	CCW	BD AND LD

KEY:
CW— CLOCKWISE
CCW— COUNTERCLOCKWISE
B— BIG TOE
L— LITTLE TOE
U— UP
D— DOWN

Comments

In this design only one arm is controlled (three functions, two directions each).
Otherwise a combination of EMG- and toe-control must be used.
All other actuation combinations are inhibited.
No speed control is possible.
Modifications of the three degrees of freedom arm for cases of *shoulder disarticulation* are also similarly possible when speed control is eliminated or when LRD, LLD are used for shoulder movements.

APPENDIX III—SIMULTANEOUS ACTUATION OF TWO LIMB FUNCTIONS

The design of Table 1 of the main text can be modified to facilitate simultaneous actuation of elbow and/or grasp and/or wrist movements. These are facilitated by modifying Table 1 such that speed control is eliminated, and the actuation scheme of Table 3 is followed.

In that case, a joint actuation of BRU and LRD or of BLU and LLD is inhibited, and only BRU or BLU alone are executed. Note that Table 4 relates to a right-arm prosthesis. In the case of a left-arm prosthesis, R and L of Table 4 should be interchanged.

TABLE 3

Function		BR		BL		LR	LL
		D	U	D	U	D	D
Prehension	Close	X					
	Open		X				
Wrist	CCW			X			
	CW				X		
Elbow	Flex					X	
	Extend						X